LIMITED COHERENCE STEREO OPHTHALMOSCOPE

Field of the Invention

The present invention relates to a method and apparatus for the stereoscopic examination of, for example, the fundus of the eye, with applications in the investigation and diagnosis of diseases that affect the posterior chamber of the eyeball. The invention will be described by reference to such applications, but it is envisioned that the apparatus and method of the present invention may be used for stereoscopic imaging in other medical processes.

Background Art

Visualisation of the ocular fundus can provide important information about the state of the eye and of the body. Information concerning ocular and systemic diseases, such as glaucoma, macula degeneration, diabetes or hypertension can be gained from examination of the posterior pole of the eye. In the past, imaging of the ocular fundus has been performed by means of an ophthalmoscope, with which a direct view of the retina can be obtained. Other methods include the use of fundus cameras to obtain photographic images. However, these techniques usually require the use of mydriatic dilating drugs. The amount of light required to illuminate the fundus may also be uncomfortable for the patient.

Recent developments have resulted in the emergence of a new imaging instrument for the ophthalmologist, in which an image of the eye may be observed in real-time and captured on a television monitor or screen, during procedures such as fluorescein angiography. This instrument, known as a Scanning Laser Ophthalmoscope (SLO), first described in US Patent 4,213,678, is currently used to produce representations of the ocular fundus in two dimensions. US Patents 4,765,730, 5,268,711 and 5,430,509 describe different embodiments of the scanning laser ophthalmoscope. All utilise a laser beam or light source that is directed through the pupil and onto the retina by way of two-directional, scanning

mechanisms. Light from the laser is reflected off the retinal wall towards a photosensitive detection device. Electro-optical circuitry is employed to convert the light into synchronized signals, so that it is possible to display an image of the fundus on a television screen or monitor.

However, although the optic disc region and retinal layers have a three dimensional structure, the existing SLO technology described above does not permit stereoscopic viewing of the ocular fundus. Stereoscopic images of the ocular fundus can impart valuable information that cannot otherwise be derived from a two dimensional representation, especially in relation to the diagnosis of glaucoma. Efforts have therefore been made to create a device capable of producing three dimensional fundus images, while improving on the contrast and resolution of conventional SLO images.

Frambach, Dacey and Sadun (1992, 1993) describe a method of producing a three dimensional fundus picture during fluorescein angiography, using a modified SLO. To obtain stereoscopic data the SLO was manually moved from side to side during angiogram proceedings, much like a fundus camera is moved to enable viewing from two different positions. individual frames from the video tape were chosen from left and right perspectives to provide a three dimensional image. An alternative approach employed by the Frambach et al. involved the use of a modified Allen separator. A piece of flat glass was attached to an extended rod, which was coupled to the Allen separator, so that the glass was interposed between the eye and the SLO. The glass was then rapidly rotated to provide the left and right perspectives. The resulting frames were digitized by computer and viewed directly on a video screen. Superimposed images were formed by breaking a stereo pair down into corresponding fields and recombining them to form a single frame. LCD glasses were then used to view the left and right fields with the corresponding eye in turn.

These disclosures illustrate that achieving a stereoscopic image from a conventional scanning laser ophthalmoscope is possible. However, the methods

involved exhibit a number of disadvantages. Frambach et al.'s first method of shifting the SLO involved awkward and confusing adjustments, resulting in poor stereoscopic image quality. The second method also had the disadvantage that interference due to unwanted back reflections from the Allen separator would hinder the transmission of stereoscopic information. Unrequired scattered light would impinge on the photodetecting element, causing a decrease in the contrast and resolution of the images.

Improvements in SLO image resolution and contrast are possible if the detector receives light only from the plane of interest and not scattered light from the media of the eye. A scanning laser ophthalmoscope that could provide high resolution, high contrast images of an ocular fundus was realised with the invention of the confocal scanning laser ophthalmoscope (cSLO), such as that described in US Patents 5,170,276 and 5,071,246. The confocal SLO utilises a pinhole or slit aperture to focus the light reflected from the fundus onto a photodetecting element. The aperture is located at a plane in which the opening is conjugate with the plane of the fundus of the eye. In this way, only the light reflected from the plane of interest impinges on the photodetecting element and any light scattered or reflected from out-of-focus planes is prevented from degrading the image.

Confocal scanning laser ophthalmoscopes are also currently used to provide three dimensional information concerning the ocular fundus. The confocal aperture of the cSLO allows the user to focus precisely on specific layers of the retina. By adjusting the focal plane of the aperture, images can be captured at different levels in the fundus, to reproduce desired depth characteristics. In this way a number of "optical sections" can be produced. A computer can then be used to extract depth information, through the process of "stacking" a selection of the optical sections taken at different levels of the retina. Information regarding the third dimension can therefore be interpolated.

US Patent No. 4,900,144 (also see Optics Communications: 87(1,2): 9-14)

describes a scanning laser ophthalmoscope that employs an alternative confocal focussing arrangement. The invention can produce a three dimensional representation of an object that displays multiple reflectivity characteristics (such as the ocular fundus) through a method slightly different from the conventional confocal depth production methods described above. This US patent teaches the use of two separate confocal slit apertures and photodetecting units. The detection slits are orientated parallel to the direction in which the light, reflected from the fundus, is scanned. However, both slits are slightly displaced from the normal position: the apertures are not conjugate with the fundus of the eye. One is positioned slightly forward of the conjugate plane, while the other is placed to the rear. Owing to the positioning of the confocal apertures, the output signals from the photodetectors vary in intensity according to the unevenness of the fundus. The resulting output signals are processed electronically by division calculations, detailed in US Patent No 4,900,144, to obtain a three dimensional profile. The resultant real-time image displays the topography of the fundus through different shade levels, reflecting different retinal depth levels. Software may also be used to create three dimensional graphic patterns.

The methods described above use the depth discrimination, or axial resolution, of the confocal system. Unlike lateral resolution, axial resolution is strongly limited by two factors. Firstly, the shape and size of the laser light focus which is scanned over the retina may suffer from deformations and distortions, particularly in the direction of the optical axis. This is due to the limited useful numerical aperture of the eye and its optical imperfections. Furthermore, the axial resolution of a confocal SLO may be constrained by the size of the detection pinhole or slit. Owing to intensity limitations on the living eye it is often necessary to provide a detector aperture size that is larger than the optimal confocal pinhole in order to maintain a sufficient signal to noise ratio. Owing to these two factors, the axial resolution of a confocal system is typically thirty times less than the lateral resolution.

A further technique to produce three dimensional images of the ocular

fundus, known as scanning laser triangulation, is described by Milbocker and Reznichenko (1991). Triangulation is a method commonly used for measuring distances. Combined with a confocal aperture, this method involves synchronized scanning of a pixel of light across the fundus by way of two mirrors. The illumination and detection paths are arranged symmetrically and are defined by the two mirrors. The axial distance is measured by the displacement of the illuminated spot in the confocal plane, enabling calculation of the depth from the points above and below the average, position of the retinal wall. With this method the axial resolution is directly coupled to the lateral resolution. However, owing to the pupil size of the human eye the useful triangulation angles are small. Consequently, the axial resolution is about 10 times worse than the lateral resolution.

Interferometry is an optical method that is much better suited for distance measurements since it does not depend on the focussing or imaging qualities of optical elements as do, for example, the cornea and the lens of the eye. Interference patterns can only occur when the difference between the length of the reference arm and the object arm is shorter than the coherence length of the light source. Non-laser light sources such as bulbs, LEDs or superluminescent diodes have a coherence length of only a few micrometres. Thus, the detection of interference patterns means that the object distance is equal to the reference distance determined with an accuracy equal to the coherence length.

Popoleanu *et al. (Journal of Biomedical Optics,* Jan. 1998, vol. 3, no. 1, p. 12-20) describe an apparatus suited for transversal and longitudinal imaging of the retina using low coherence reflectometry. The light in the object arm of a fibre-based interferometer, where a superluminescent diode is the light source, propagates through a phase modulator and is scanned over the retina in a raster pattern. The light reflected from the fundus of the eye is combined with the light in the reference arm of the interferometer which length is controlled by means of moveable mirrors. Frequency sensitive detection recognizes the occurrence of interference. The strength of the interference signal is used by a frame grabber to

form an image of features that are situated in a thin layer in the back of the eye. The axial position of this layer is controlled by the movable mirrors in the reference arm while the thickness of the layer is determined by the coherence length of the light source. In much the same way as with a confocal scanning laser ophthalmoscope a three dimensional image can be computed from a number of optical sections (transversal images) at different axial positions. Alternatively, scans in only one lateral and the axial direction (x-z-scans) may be carried out to record longitudinal images. The images acquired with this apparatus have a high depth resolution, about ten times better than a confocal scanning laser ophthalmoscope. However, the length of the object arm of the interferometer varies with the scan and/or the focal position as well as with the position and the size of the individual eye which is examined. It is therefore impractical or impossible to obtain quantitative data for the measurement of absolute distances in the eye ball. Furthermore, with the described lay-out a reduction in recording time is difficult due to bandwidth limitations of the phase modulation and the speed of the galvo-scanners.

A dual beam system as described by Baumgartner *et al. (Journal of Biomedical Optics*, Jan. 1998, vol. 3, no. 1, 45-54) overcomes one of the aforesaid problems. An external interferometer produces a beam with two coaxial components which have a path difference of twice the difference in the arm length of the interferometer. This beam is guided onto the eye, where parts of the beam are reflected from the cornea - which acts as a reference surface - and other parts are reflected from the fundus of the eye. If the optical distance between the cornea and a certain fundus feature matches the difference in length between the interferometer arms (within the accuracy of the coherence length of the light source) interference signals are detected. It is then straightforward to get a readout for the distance in the eye ball from the positions of the two interferometer mirrors. In order to achieve good signal-to-noise ratios for the interference detection a sub-component of the illumination beam has to be focused on the cornea while the rest of the beam has to enter the pupil of the eye in a more or less collimated state. In this configuration the focused component travels through

air only whereas the collimated component has to travel through the dispersion causing ocular medium also. It is demonstrated that a dispersion compensating element can therefore be used advantageously and an axial resolution of only a few micrometres can be achieved. To split the illuminating beam into a focused and a collimated component Baumgartner et al. use a special Fresnel lens-like diffractive optical element which is placed in front of the eye. However, the use of this non-standard element can cause unwanted back reflections and transmission losses. Lateral scanning of the beam, which is necessary to acquire data not only for one point but for a line or an area of the retina, is also severely restricted.

In order to detect the occurrence of interference a time variation in the interference pattern is necessary. In this lay-out the Doppler-shift caused by the moving reference mirror creates this time variation. it is therefore not possible to record a limited coherence image of a layer of the retina for a fixed axial position of this layer.

Accordingly, there remains a need to provide apparatus able to be adapted, at least in an embodiment, as a limited coherence scanning ophthalmoscope capable of acquiring images with a high depth resolution and of quantifying the 3D-morphology of the back of the eye, which preferably is not restricted by any of the aforementioned limitations.

It is therefore an object of the present invention, at least in one or more preferred embodiments, to provide an improved method and apparatus for producing a high contrast, three dimensional representation of a scanned object, based on both the high lateral resolution of the reflection characteristics of that object and the high axial resolution of the limited coherence reflectometry.

It is a further object of the present invention, at least in an advantageous application, to achieve the above object with a novel limited coherence scanning ophthalmoscope design that incorporates the use of additional beam splitting, focussing and beam combining components to gain information and quantitative

topographic data about the third dimension.

Summary of the Invention

In a first aspect, the invention provides an imaging apparatus for the three dimensional imaging and/or measurement of a surface including:

first beam modifying means for modifying an incident beam of short coherence length light to form a modified beam of first and second components having a mutual path difference and being capable of producing a detectable interference;

beamsplitting means for splitting said modified beam into first and second beams;

second beam modifying means for modifying the properties of at least one of said first and second beams;

recombining means for thereafter recombining said first and second beams;

means for directing said recombined first and second beams towards said surface and scanning them across the surface; and

means for monitoring the first and second beams after reflection and detecting interference of the reflected first and second beams.

Preferably, in its first aspect, the invention further includes steering means to vary a nodal point of the scanned first and second beams.

Preferably, the first first beam after said modification of the properties of at least one of the first and second beams is focused on a position in front of the surface for reflection at said position. The second beam may be a collimated beam at the scanning means, for being focussed onto the surface.

Advantageously, the scanning means is arranged for scanning in at least two dimensions.

In an advantageous application, the apparatus is for imaging and/or

measuring a surface comprising an ocular fundus, eg the incident beam is a laser beam, and the apparatus functions in use as a scanning laser ophthalmoscope. In this case, the first and second beams may respectively be a focussed beam arranged to be at least partially reflected from the cornea of an eye, and a collimated beam for being focussed by the eye onto its fundus for reflection thereby.

According to a second aspect of the present invention, there is provided an imaging apparatus for the three dimensional imaging and measurement of a surface including:

a beam source for providing a beam of short coherence length light;

a first beamsplitter for splitting said beam into first and second components of short coherence length light;

means for producing a path difference between said first and second components;

a second beamsplitter for splitting said beam into first and second beams;

beam modifying means for modifying the properties of at least one of said first and second beams;

recombining means for recombining said first and second beams;

focussing means for focussing said recombined first and second beams;

first and second beam scanners for scanning said recombined first and second beams in first and second directions;

beam steering means for creating a triangulation base by directing said recombined first and second beams onto said surface from a first and a second position and reflecting said recombined first and second beams therefrom;

a third beamsplitter for splitting said reflected first and second beams; and

a detector for detecting the interference of said reflected first and second beams.

In a third aspect, the invention provides apparatus for visualising the ocular fundus of an eye and providing three dimensional topological data of said fundus including:

a light source for producing a beam of short coherence length;

an interferometer for dividing said beam into sub-components with a defined path difference;

modulation means for modulating at least one of the said sub-components;

beam shaping means for shaping said beam and/or said sub-components;

polarisation influencing means for controlling and changing the polarisation state of said beam and/or said sub-components;

a first beamsplitter and recombining means for splitting and re-combing said sub-components;

first focussing means for focussing said sub-components;

a second beamsplitter for splitting the sub-components;

second focussing means for further focussing at least one sub-component;

first scanning means for scanning the sub-components in a first direction;

second scanning means for rescanning the sub-components in a second direction substantially perpendicular to said first direction and thereby converted into a raster pattern;

a beam steerer for creating a triangulation base and directing said sub-components onto said fundus and the cornea of said eye respectively;

light detecting means for detecting said interference pattern after re-combining reflected light from said fundus and said cornea;

signal processing means for processing signals from said light detecting means; and

display means for receiving said processed signals and displaying an image of said fundus.

Preferably, in use, light travels from said beam source to said surface along an input path and, after reflection from said surface, an output path wherein said input and output paths are identical at least in part.

Preferably said apparatus includes interferometric means having at least one optical arm with adjustable mirror means, for example, the interferometric means may include a first beamsplitter and two mirrors wherein at least one of the mirrors is movable and position controllable.

Preferably the apparatus includes modulation means for modulating at least one of the first and second beams, eg the phase difference between the first and second components.

The modulation means may include an electro-optic phase modulator, or alternatively a fibre based phase modulator.

The scanning means may include a mirror on a resonant scanner, or a rotating polygon mirror, a mirror on a galvanometer motor, or an acousto-optic deflector, and/or a mirror mounted on a scanning galvanometer motor.

The beam steering means may include a pair of toggling mirrors that toggle every alternate frame or half frame to image the surface from two different positions, with substantially overlapping imaging areas, such that a triangulation base can be created.

The apparatus preferably includes signal processing means, eg. a computer with a video signal capture facility, and display means such as, eg. a computer monitor.

Preferably the apparatus includes imaging analyzing means to obtain three dimensional topological data of the surface. Such image analyzing means advantageously includes computation means for identifying image features and carrying out length measurements. The computation means may typically include

computer hardware and/or software. Such software may be conveniently include an image and data processing program that is capable of recognizing and extracting image features, carrying out distance measurements and producing three dimensional topological data.

In a fourth aspect, the invention provides a method for the three dimensional imaging and/or measurement of a surface including:

modifying an incident beam of short coherence length light to form a modified beam of first and second components having a mutual path difference and being capable of producing a detectable interference;

splitting said modified beam into first and second beams;

modifying the properties of at least one of said first and second beams and thereafter recombining said first and second beams;

directing said recombined first and second beams towards said surface and scanning them across the surface; and

monitoring the first and second beams after reflection and detecting interference of the reflected first and second beams.

The invention further provides, in a fifth aspect, a method for scanning a surface with light beams of short coherence length to thereby produce an image and three dimensional topological data of said surface, including:

directing light beams of short coherence length along an input path including:

polarising said beam;

dividing said beam into two sub-components with a defined path difference;

modulating at least one of said sub components;

splitting said sub-components;

focussing said sub-components;

re-combining said sub-components;

scanning said sub-components in a first direction;

scanning said sub-components in a second direction different from

said first direction;

directing said sub-components through a beam steerer to provide a triangulation base by impinging said sub-components onto said surface from two different positions;

reflecting said sub-components onto said surface;

whereby reflected light from said surface traverses an output path identical at least in part to said input path, including splitting said sub-components and directing a portion of the split sub-components through an aperture means towards detecting means coupled to signal processing means and display means, whereby the resultant image can be viewed and three dimensional topological data of said surface can be obtained.

Preferred, advantageous and optional features of the apparatus as described above are where appropriate also applicable as preferred, advantageous or optional steps of the method of the fourth or fifth aspect of the invention.

Brief Description of the Drawings

In order that the invention be more fully ascertained, some preferred embodiments will be described, by way of example, with reference to the accompanying drawings, in which:

Figure 1 is a schematic diagram of a first preferred embodiment of the present invention; and

Figure 2 illustrates the process of data and image acquisition.

Preferred Embodiments

Referring to Figure 1, a preferred embodiment of the present invention utilises a superluminescent diode source 1, or any other suitable source of light with a short coherence length including collimating optics, which emits a light

beam 2. Alternatively, two or more light sources may be utilised to produce beam 2. This beam is directed through a polariser 3 onto a first beamsplitter 4 which forms part of an interferometric set-up 9. Components 5 and 6 of beam 2 are impinged onto mirrors 8 and 9 respectively, and reflected back to first beamsplitter 4. Mirror 8 is movable along the axial direction and the position of mirror 8 is computer controlled. Component 6 passes through the high fixed frequency phase modulator 7 which continuously changes the phase but not the intensity or the polarisation state of component 6. Alternatively, the Doppler effect may be used to create a frequency shift in component 6 when moving the mirror 8.

Beamsplitter 4 recombines components 5 and 6 to form beam 10 which consists of two sub-components with a difference in path length determined by the positions of mirrors 8 and 9. The beam 10 passes through half-wave plate 11, which rotates the polarisation direction of the linearly polarised beam 10 and determines the relative intensities of the components 14 and 15. Alternatively, a variable retarder in conjunction with a quarter wave plate can replace half wave plate 11.

Beam 10 is then guided (through beamsplitter 12-see below) onto a second beamsplitter in the form of polarizing beamsplitter 13, which produces the two components, ie first and second beams 14 and 15 with perpendicular polarisation directions. Component 14 is impinged off mirror 17 towards lens 18 to produce a non-collimated beam in such a way that component 14 will finally be focused on the cornea of the eye 25. Alternatively, a combination of lenses and/or flat and/or curved mirrors can replace lens 18. Component 15 passes through beamsplitter 16, and a portion of component 15 reaches the beamsplitter 19, preferably a Thompson prism. Beamsplitter 19 re-combines components 14 and 15 to form beam 20. As mentioned, component 14 is focused on the cornea of eye 25, whereas component 15 is collimated.

The beam 20 is directed towards the focussing unit 21, consisting of a lens or combinations of lenses and mirrors.

After leaving the focussing unit 21 the beam is guided through the scanning unit 22 where a scan pattern is produced to image areas of interest of the eye. Within scanning unit 22 a beam may be guided onto a resonant scanner or a rapidly rotating multiple facet mirror, which acts as a horizontal scanner, and subsequently onto a small curved mirror , which shapes the beam into a horizontal line. The beam may then travel to a vertical scanner such as a galvanometer controlled mirror.

After leaving the scanning unit 22 the beam 20 passes through a beam steerer in the form of beam steering unit 23, which allows for moving the nodal point of the scan pattern to different positions within the pupil. Preferably, the beam steering unit 23 consists of a pair of toggling mirrors. These mirrors are positioned so that each directs the beam 20 onto the surface to be imaged from two slightly different directions and positions, with substantially overlapping imaging areas. They preferably toggle every alternate frame, such that visual information is received from the right and left perspectives in alternate frames. Alternatively, the two toggling mirrors could be substituted with a single mirror that can change position preferably every second frame or half frame, or a galvanometer mounted prism or glass plate that is capable of imaging from the left and right perspective every half frame.

After leaving the beam steering unit 23 the light is reflected off a large curved mirror 24 before entering the eye 25. Component 14 of the beam 20 is focussed onto the cornea and is reflected off it to provide a reference signal. Component 15 of beam 20 enters the eye through the pupil and passes through the eye's internal structure to reach the retina at the back of the fundus. The light is reflected off the retinal layers and exits through the pupil.

It will be appreciated that by forming component beams 14, 15 before scanner 22, the scanner is able to scan the recombined beam, thus enhancing flexibility in lateral scanning. The prior complex diffractive element in front of the eye is also avoided.

The reflected beam 26, including reflections of incident components 14, 15, traverses the same output path as the incident beam 20. Beamsplitter 19 divides beam 26 into a component that is reflected from the cornea and a component reflected from the fundus. The latter component reaches the non-polarising beamsplitter 16 where parts of the light reflected from the fundus form beam 27. The intensity of beam 27 is measured by photodetector 28, preferably a photomultiplier tube or an avalanche photodiode, to form the standard scanning ophthalmoscopic image of the fundus of the eye. Alternatively, focussing optics and/or an aperture may be placed in front of the photodetector, or optional photodetector 32 and/or 33 may be used to detect the standard scanning ophthalmoscopic image which is offset by the reflected intensity of the cornea. Alternatively, half wave plate 11 may be adjusted so that component 14 vanishes and photodetector 32 and/or 33 detects the standard scanning ophthalmoscopic image. Beamsplitter 13 re-combines the light which passes through beamsplitter 16 with light which leaves beamsplitter 19, passes through lens 18 and is reflected off mirror 17 to reform beam 26.

Upon reaching the third beamsplitter in the form of nonpolarizing beamsplitter 12, part of the light is reflected by the beamsplitter towards the polarising beamsplitter 29 which is rotated by 45°. Both the beams 30 and 31, which are polarised perpendicular to each other, contain reflected light from both the cornea and the fundus of the eye 25. Interference in both the beams 31 and 32 will occur when the path difference of the light reflected from the cornea and the light reflected from the fundus equals the optical distance between the cornea and the fundus within the tolerances of the coherence length of the light source 1. Since the path difference of the two subcomponents of beams 10 and 20 is set by the positions of the mirrors 8 and 9, the distance from the fundus feature which is imaged to the cornea can be measured by determining the positions of the mirrors 8 and 9 when interference is established.

The intensities of the beams 30 and 31 are detected using the photodetectors 32 and 33, by preference photomultiplier tubes or avalanche

photodiodes. Alternatively, focussing optics and/or apertures may be placed in front of the photodetectors. Signal processing means 34 and 35 compare the signals from the photodetectors 32 and 33 with the frequency of the phase modulator 7 to recognise interference patterns. Alternatively, the signals of the photodetectors 32 and 33 can be compared to a frequency resulting from the Doppler shift when moving mirror 8. Alternatively, the signals of the photodetectors 32 and 33 may be subtracted from each other to reveal interference patterns.

The outputs of the photodetector 28 and the signal processing means 34 and 35 are then sent to the computer 36. An imaging board in computer 36 and appropriate electronic hardware and software convert these outputs and information on the position of the mirror 8 into images, including three dimensional topological representations of the fundus of the eye. The images are displayed by computer monitor 37. It is then possible to calculate three dimensional distances between features chosen by the computer operator from those images.

Referring to Figure 2, a preferred procedure of the present invention utilises a beam consisting of two sub-components with a fixed path difference. The nodal point of the scanning pattern is located at the cornea at position A. Lateral scanning takes place and a standard ophthalmoscopic image is recorded. Simultaneously, the points and/or areas where interference occurs are marked in a different colour on the same image. In doing so, the classification of these features which cause the interference is facilitated. The image is stored and the nodal point of the scanning pattern is moved to position B. Again, lateral scanning takes place and the points and/or areas of interference overlaid onto a standard ophthalmoscopic image are recorded. After storing the image the path difference between the sub-components of the illumination beam is changed by a certain amount (step width) and images at the positions A and B are acquired in the same manner as described above. The process of changing the path difference between the sub-components of the illumination beam and the acquisition of images at the positions A and B continues until the full three dimensional region of

interest of the fundus of the eye is covered.

In order to determine the distance of the feature D from the reference line AB computer software is used to detect those images in the stacks of images recorded from positions A and B respectively where the feature D is marked as causing interference. The path difference between the sub-components associated with these images is a measure of the distances AD and BD, respectively. The distance AB is controlled by the positions of certain optical elements of the apparatus shown in Figure 1 and therefore also known. As a result, the triangle ABD is completely determined without measuring any angles. Triangulation methods can now be used to calculate the length of any line, including the height h (the height of triangle ABD along CD) and the value of any angle of this triangle.

Further information such as the value of the refractive index can be obtained from angular measurements. The nodal point of the scanning pattern may be placed at position C on the cornea, halfway between positions A and B. The horizontal scan angle under which the feature D is imaged corresponds to a certain position of the horizontal scanner of the apparatus shown in Figure 1 and can be detected. The angle β, which describes the direction a beam has to travel from position C to reach feature D, can be calculated using trigonometric methods. The refractive index is then given as the quotient of sin and sin β.

Modification within the spirit and scope of the aforementioned invention may be readily effected by a person skilled in the art. Other alternative embodiments would involve the use of fibre-optic based light delivery systems including fibre-optic based components for beam splitting and re-combing and phase modulating. The position of the beam splitting, focussing and re-combining unit which is used to produce a reference beam focused on the cornea, may be changed within the lay-out of the complete set-up. The process of image and data acquisition may also be altered, for example, axial scanning may be carried out for a fixed lateral position or in conjunction with a line scan. It is to be understood,

therefore, that this invention is not limited to the particular embodiments and methods described by way of example hereinabove.